A METHOD TO IMPROVE THE CALCULATION OF KNEE CONFIGURATION ANGLES IN CLINICAL AND SPORT BIOMECHANICS

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A METHOD TO IMPROVE THE CALCULATION OF KNEE CONFIGURATION ANGLES IN CLINICAL AND SPORT BIOMECHANICS

(Thesis Abstract)

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Knee injuries are quite common in sports activities. Biomechanical analyses seek to improve understanding of the mechanisms that produce injury, and to find ways to reduce the incidence of injury. The calculation of knee torques and knee configurations requires the establishment of three-dimensional reference frames attached to the thigh and to the shank. Most clinical biomechanics researchers use the methods proposed by Kadaba et al. (1990) and by Davis and al. (1991). The calculation of the resultant knee joint torques and of the knee joint deformation is hampered by an important methodological problem. The problem is that usually no distinction is made between knee configuration angles and knee deformation angles. The primary purpose of this study was to look for a solution to this problem.

Ten male subjects and four female subjects were recruited to participate in the investigation. The subjects performed three types of tests. In the first type the subject performed slow flexions and slow extensions of the right knee in unloaded conditions. In the second type of test they performed a series of 4 straight runs. In the third type of test they performed a series of 4 trials in which they ran forward and executed a cutting maneuver to the left. The trials of types 2 and 3 were given in random order. Three-dimensional leg segment orientations and joint angles were calculated from the landmark location data.

The results support the measurement of angles in unloaded trials to provide adjustments for the raw ab/adduction angles of loaded activities. On the other hand, such an approach is not currently possible for the internal/external rotation angles due to the large intra-subject variability of the internal/external rotation angles in the unloaded trials. The adjusted ab/adduction angles generally reached less extreme values than the unadjusted ones. The standard Kadaba/Davis methods do not include such adjustments. Therefore, the ab/adduction angles calculated with those methods are inflated.

In summary, the present project demonstrated a new method for improving the calculation of knee configuration angles. These results can be applied in both clinical and sport biomechanics in ways that will ultimately be of benefit in the future study and treatment of knee injuries.

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INTRODUCTION

Knee injuries are quite common in sports activities. For instance, in men's and women's NCAA basketball and soccer they occur at a rate of between 0.7 and 1.6 per 1000 practices or games, and amount to between 12% and 19% of all injuries (Arendt & Dick, 1995).

Biomechanical analyses seek to improve understanding of the mechanisms that produce injury, and to find ways to reduce the incidence of injury. These biomechanical studies usually include the analysis of resultant knee joint torques and knee joint configurations. One of the goals is to understand the relationship between the joint torques exerted by the thigh on the shank through muscles, ligaments and bones, and the deformation of the knee joint while subjected to these torques. An important aspect of the knee joint deformation is the changes in the knee configuration angles in the directions of varus/valgus and internal/external rotation.

The calculation of knee torques and knee configurations requires the establishment of three-dimensional (3D) reference frames attached to the thigh and to the shank. Each segment has an axis that points along the sequent's longitudinal direction, and two transverse axes perpendicular to it. The calculation of the direction of the longitudinal axis is quite straight-forward, but the transverse axes present greater difficulties.

Most clinical biomechanics researchers use the methods proposed by Kadaba et al. (1990) and by Davis and al. (1991). These two methods are quite similar, and from here on they will be referred to collectively as the Kadaba/Davis methods.

In the Kadaba/Davis methods the longitudinal axis of the thigh is defined as a line pointing from the knee joint to the hip joint. The knee joint is defined as the midpoint between two reflective markers positioned at the medial and lateral epicondyles of the femur. The location of the hip joint is calculated from the locations of the anterior superior iliac spines

(ASIS) and of the sacrum or the posterior superior iliac spines (PSIS) using mathematical models based on pelvic radiography studies.

In the Kadaba/Davis methods the mediolateral axis of the thigh is defined as perpendicular to the longitudinal axis, and contained in the plane defined by the hip joint and the two epicondyles of the femur. The anteroposterior axis is perpendicular to the longitudinal and mediolateral axes.

The letter labels for the three axes and the choices for the axes' positive and negative directions have varied widely in the field applications of the Kadaba/Davis methods (Kadaba et al., 1990; Davis et al., 1991; Besier et al., 2001; Malinzak et al., 2001; Ferber et al., 2003; McLean et al., 2004; Pollard et al., 2004; MacLean et al., 2006; Chow et al., 2009; Uygur et al., 2009). For the sake of clarity, from here on the mediolateral, anteroposterior and longitudinal axes will be labeled as X, Y and Z, respectively, and the positive directions of these axes will point in the lateral, anterior and proximal directions, respectively.

The configuration of the knee joint at any instant is usually defined as the 3D orientation of the shank relative to the thigh at that instant. This is normally expressed in terms of Cardan angles. The Cardan angles α, β and γ represent the three successive rotations about the X, Y and Z axes of the thigh that would make these three axes coincide with the X, Y and Z axes of the shank. These three rotations are then usually considered to represent, respectively, the angles of flexion/extension, varus/valgus, and internal/external rotation of the knee joint at that instant. The angles of varus/valgus and internal/external rotation of the knee might be interpreted as representing the knee deformations produced in these directions by external load. However, this is not necessarily the case, as the unloaded knee may already have varus/valgus and internal/external rotation angles. This problem seems to have an easy solution: Subtract the angles of the unloaded condition from those of the loaded condition. However the issue is more complex, since the varus/valgus and internal/external rotations of the knee in the unloaded condition should be expected to vary throughout the range of knee flexion/extension.

A solution to the problem is to perform an unloaded trial that covers the entire range of motion of the knee to establish the dependence of the β and γ Cardan angles upon the simple knee angle of the unloaded condition (defined as the angle between a vector pointing from knee to ankle and a vector pointing from knee to hip joint), and then subtract these β and γ angles from those of loaded trials to calculate the β and γ deformations of the knee that were due to the load.

METHODS

General Procedures

Ten male subjects (height: 1.87 m (SD, 0.11 m); mass: 90 kg (SD, 14 kg)) and four female subjects (height: 1.67 m (SD, 0.11 m); mass: 69 kg (SD, 19 kg)) participated in the investigation. All the subjects were varsity basketball players. Permission to conduct the study was obtained from the Indiana University – Bloomington Campus Committee for the Protection of Human Subjects. Informed consent was obtained from the participants before the study began. At the time of the study all participants were healthy, with no complaints of hip, knee, ankle or other musculoskeletal injuries.

Data collection took place at the Indiana University – Bloomington Biomechanics Laboratory. The subjects performed three types of tests: (1) In the first type the subject performed three slow flexions and three slow extensions of the right knee in unloaded conditions; (2) in the second type of test the subjects performed a series of 4 straight running trials in which they made a short forward run (about 12 meters) at a moderate target speed of about 5-6 m/s, stepping with the right leg on a force platform about 2/3 of the way into the run; (3) in the third type of test the subjects performed a series of 4 trials in which they ran forward about 8 meters at a moderate target speed of about 5-6 m/s, stepped on the force platform with the right leg, and executed a cutting maneuver (a change of direction attempting to step onto a lane 0.30 meters wide set at a 45° angle to the left relative to the initial part of the run). The trials of types 2 and 3 were executed in random order. The motions of selected anatomical landmarks of the subjects were recorded with an automatic motion capture system. Three-dimensional leg segment orientations and joint angles were calculated from the landmark location data.

Recording Techniques

Each trial was recorded with a Vicon 370 3D motion analysis system (Oxford Metrics, Ltd., Oxford, United Kingdom). Six cameras captured the motions of spherical reflective markers attached to the subject, at a sampling rate of 60 Hz. The Vicon motion analysis system calculated 3D coordinates for the markers. The instants of right foot landing and takeoff in the straight running and in the cutting maneuver were determined using an AMTI Model OR6-7-1000 force platform embedded in the floor (Advanced Mechanical Technology, Inc., Watertown, MA) synchronized with the Vicon and gathering data at a sampling rate of 600 Hz.

Ten spherical reflective markers (diameter = 25 mm) were attached to the subjects' skin: surfaces of the left and right anterior superior iliac spines (ASIS); surfaces of the left and right posterior superior iliac spines (PSIS); medial and lateral surfaces of the femoral epicondyles of the right leg ("knee markers"); medial and lateral malleoli of the right leg ("ankle markers").

Analysis of the Recorded Data

Calculation of marker locations

The 3D coordinates of all reflective markers were expressed in terms of an inertial reference frame $\mathbf{R_0}$. The origin of reference frame $\mathbf{R_0}$ was at ground level, at the center of the force platform. Its axes were defined by vectors $\mathbf{X_0}$, $\mathbf{Y_0}$, and $\mathbf{Z_0}$. $\mathbf{Z_0}$ was vertical, and pointed upward; $\mathbf{Y_0}$ was horizontal and pointed forward along the long side of the force platform, in the direction faced by the subject at the start of the run; $\mathbf{X_0}$ was horizontal, and pointed toward the right.

Calculation of anatomical landmarks

The positions of the hip joint centers were estimated from the positions of the ASIS and PSIS marker centers (Meyer, 2005). First, the positions of the ASIS bony landmarks were estimated from the positions of the ASIS marker centers, the distance from each marker center to the corresponding ASIS bony landmark, and a convergence angle in the transverse plane of the pelvis. The distance from the ASIS marker center to the ASIS bony landmark was the sum of the 15.7 mm distance from the marker center to the skin and 8 mm of estimated tissue thickness (ASPECT Report, cited by Bush & Gutowski, 2003). The convergence angles were computed from the distance from marker centers to skin, the distance between the ASIS markers, and the average ratio between the distance between markers and the distance of their points of attachment to the skin (1.025) according to Wetherington (2009). Pelvic width was defined as the distance between the ASIS bony landmarks. The right hip joint was estimated to be located at distances of 36%, 22% and 30% of pelvic width laterally, posteriorly and caudally, relative to the midpoint between the ASIS bony landmarks (Bell, Brand, & Pedersen, 1989; Seidel, Marchinda, Dijkers & Soutas-Little, 1995).

The right knee was calculated as the midpoint between the medial and lateral knee markers, and the right ankle as the midpoint between the medial and lateral ankle markers.

Smoothing of landmark locations

Quintic spline functions (Woltring, 1986) were fit to the time-dependent $\mathbf{X_0}$, $\mathbf{Y_0}$, and $\mathbf{Z_0}$ coordinates of each body landmark with a smoothing factor equivalent to a 15 Hz digital filter.

Calculation of hip velocities

Hip velocity was used as a surrogate for center of mass velocity. Average X and Y components of hip velocity were calculated for the airborne phases immediately before and after the analyzed support phase.

Calculation of segmental reference frames

Three non-inertial right-handed orthogonal reference frames were defined (for the pelvis, and for the thigh and shank of the right leg). These reference frames will be described next.

Pelvis reference frame. Reference frame $\mathbf{R_P}$ was attached to the pelvis. $\mathbf{X_P}$ was a vector pointing from the left ASIS marker toward the right ASIS marker; $\mathbf{Z_P}$ was be the cross product of $\mathbf{X_P}$ and a vector pointing from the midpoint of the two PSIS markers toward the midpoint of the two ASIS markers; $\mathbf{Y_P}$ was be the cross product of and $\mathbf{Z_P}$ and $\mathbf{X_P}$.

Thigh reference frame. Reference frame $\mathbf{R_T}$ was attached to the right thigh. $\mathbf{Z_T}$ was a vector pointing from the right knee toward the right hip joint; $\mathbf{Y_T}$ was the cross product of $\mathbf{Z_T}$ and a vector pointing from the medial knee marker toward the lateral knee marker; $\mathbf{X_T}$ was the cross product of $\mathbf{Y_T}$ and $\mathbf{Z_T}$.

Shank reference frame. Reference frame $\mathbf{R_S}$ was attached to the right shank. $\mathbf{Z_S}$ was a vector pointing from the right ankle toward the right knee; $\mathbf{Y_S}$ was the cross product of $\mathbf{Z_S}$ and a vector pointing from the medial ankle marker toward the lateral ankle marker; $\mathbf{X_S}$ was the cross product of $\mathbf{Y_S}$ and $\mathbf{Z_S}$.

The 3D orientations of the three axes of $\mathbf{R_T}$ and $\mathbf{R_S}$ were calculated for all times of the unloaded trials. The angle κ between $\mathbf{Z_T}$ and a vector pointing from the knee to the ankle (the "simple knee angle") was calculated for all instants of the unloaded trial.

Calculation of joint angles for the knee

The knee configuration was calculated for every instant of the unloaded trial. This was expressed in terms of the three Cardan angles (α, β and γ) that would be needed to rotate $\mathbf{R_T}$ into an orientation identical to that of $\mathbf{R_S}$. The rotations were in the order X-Y-Z, and the second and third rotations were about axes displaced by the previous rotation(s).

The α, β and γ angles were then plotted against the κ angle throughout the unloaded trial, and separate 6^{th} degree polynomials were fitted to each of these three relationships. These functions described the relationships between the κ angle and the α, β and γ angles in the unloaded trial. (The supplementary of the κ angle was similar but generally not identical to the α angle.) The equations of these three polynomials were output into a computer file.

Calculation of joint angles for the knee

In the loaded trials, the orientation of the shank relative to the thigh was also expressed using Cardan angles. The three sequential angles were labeled flexion/extension, varus/valgus, and internal/external rotation. Positive angles corresponded to extension, varus rotation, and internal rotation at the knee.

Calculation of the knee deformation angles in the loaded trials

The κ angle and the α, β and γ angles of the shank relative to the thigh were calculated for every instant of the loaded conditions. The κ angle was then used with the formulas of the three polynomials obtained from the unloaded trial to calculate for each instant of the loaded trials the α, β and γ angles that would have occurred in unloaded conditions. The predicted β and γ angles of the unloaded condition were then subtracted from the β and γ angles of the loaded condition to calculate, respectively, the deformations of the knee joint in the directions of varus/valgus and internal/external rotation during the loaded trial.

The knee deformation angles obtained with the new method proposed in this project were compared with the values that would have been obtained using the standard Kadaba/Davis methods.

Statistically significant differences between straight runs and cutting runs and between adjusted and non-adjusted loaded knee angle data were tested using paired t-tests ($P < 0.05$).

RESULTS AND DISCUSSION

The occlusion of reflective markers made it impossible to analyze some of the trials. For the unloaded condition, all analyzable trials were used in the project. For the loaded conditions, two trials were selected arbitrarily for each individual: one straight run and one cutting run.

Table 1 shows the X and Y components of the mid-hip velocities of men and women before and after contact with the force platform in the straight and cutting runs.

Straight runs				
	Before support		After support	
	X	Y	X	Y
Men	0.0 ± 0.2 *	6.1 ± 0.5 *	0.0 ± 0.1	6.0 ± 0.8
Women	-0.1 ± 0.5	5.9 ± 0.7	0.0 ± 0.1	5.9 ± 0.8
Cutting runs				
	Before support		After support	
	X	Y	X	Y
Men	-0.8 ± 0.3	5.5 ± 0.5	-2.0 ± 0.3	5.0 ± 0.4
Women	-0.7 ± 0.7	4.7 ± 0.6	-2.2 ± 0.4	4.4 ± 0.7

Table 1. Mid-hip velocities of men and women before and after contact with the force platform in the straight and cutting runs. Data represent N = 10 for men and N = 4 for women; * is N = 9.

In the cutting runs, the subjects had less forward velocity than in the straight runs (paired t-test, $P < 0.001$) and a clear velocity component toward the left. Thus, before the foot was planted on the ground the path of the mid-hip was at an 8° angle toward the left for both men and women; after the support, the path of the mid-hip was at a 22° angle toward the left for the men, and 27° for the women. These results indicate that the subjects were anticipating the cutting maneuver, and prepared for it before the foot was planted on the ground. Because of this, the change in the direction of travel during the support was only 14° for the men and 19° for the women. Consequently, the change in direction during the support in the cutting trials was much smaller than planned. Because of this, the loads put on the support leg were probably smaller than they would have been in a support producing a more marked change of direction. Still, the cutting trials should be expected to produce larger deformations of the right knee than the straight runs, which showed essentially no change of direction during the support.

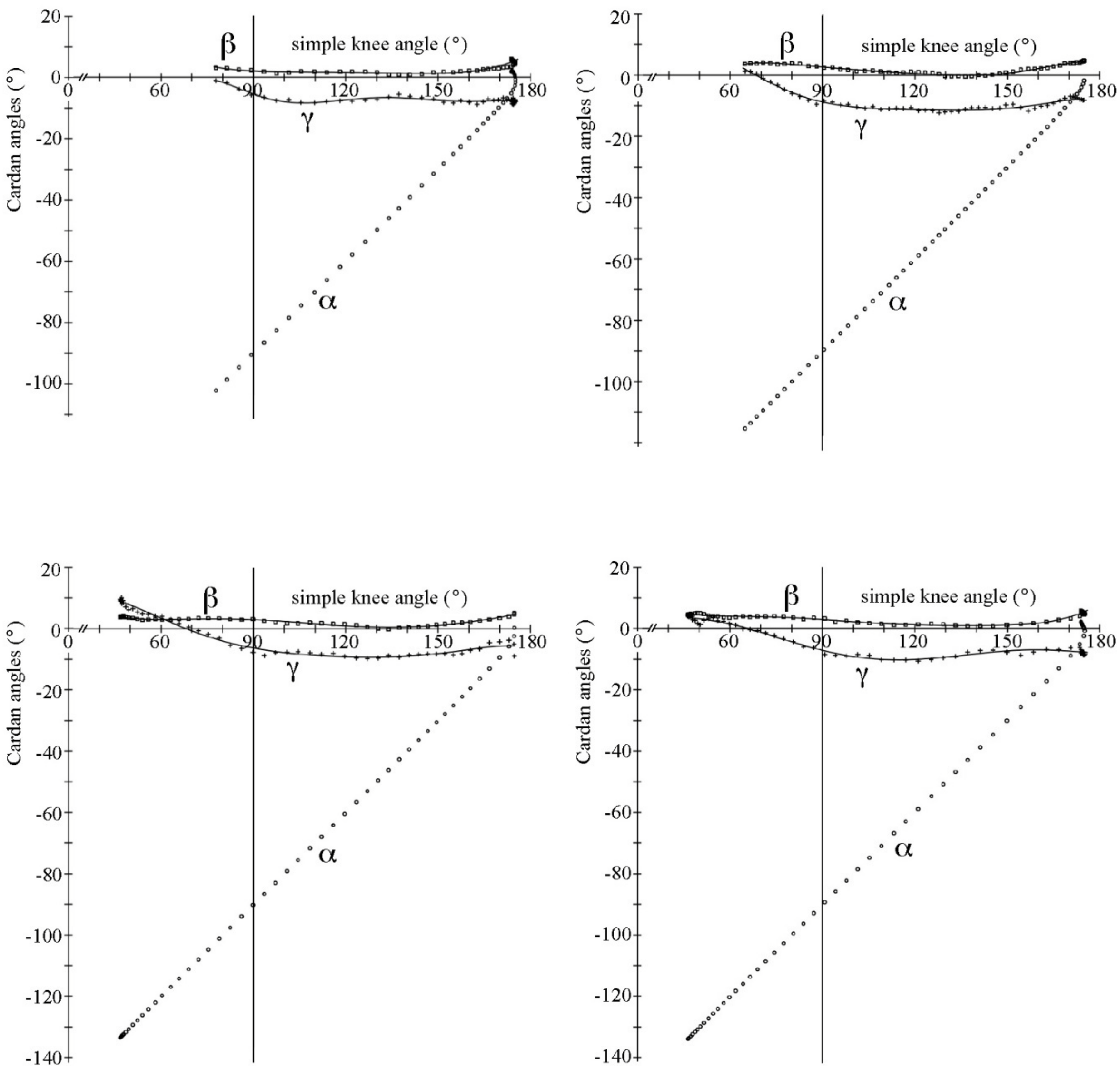

Figure 1. Relationships of α, β and γ with κ in the four available trials of a typical subject.

Figure 1 shows the relationships of α, β and γ with κ in the four available unloaded trials of a typical subject, and the 6th degree polynomials fitted to the β and γ angles.

Figure 2 shows the polynomials of the β versus κ relationships in the four unloaded trials of the typical subject shown in Figure 1. Figure 3 shows the polynomials of the γ versus κ relationships in the same trials.

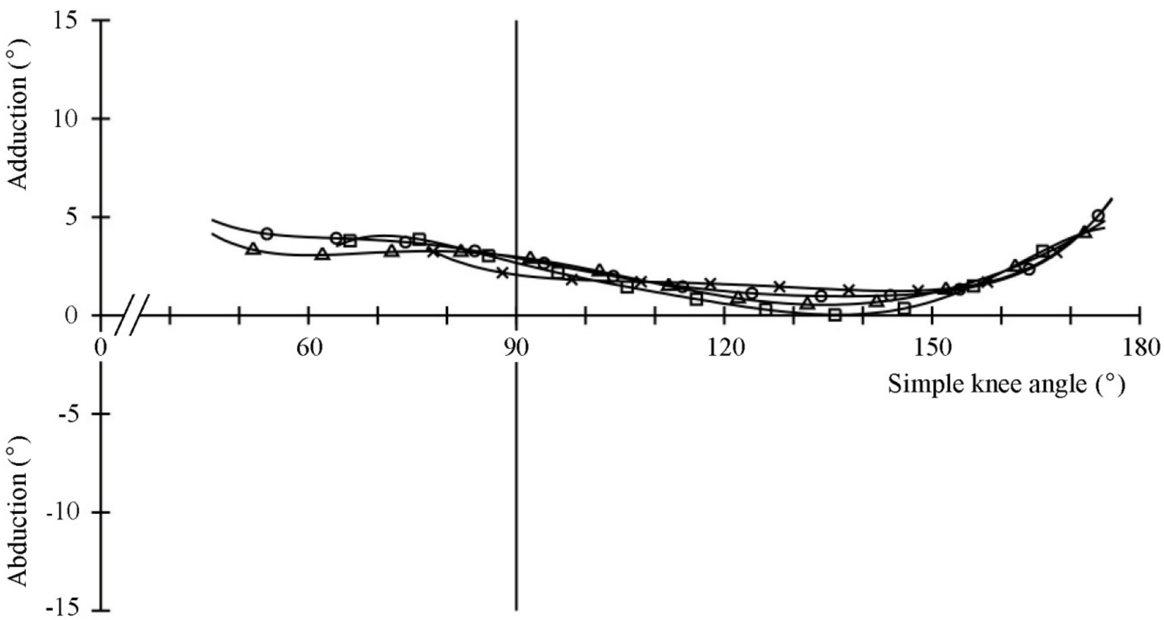

Figure 2. Polynomials of the β versus κ relationship in the four unloaded trials of a typical subject.

To evaluate the variation between unloaded trials, standard deviations were calculated for both the β and γ angles at 10° intervals of the κ angles. This was done within the range of κ angles for which data from all trials were available. The average and maximum values of the standard deviations were obtained for each subject.

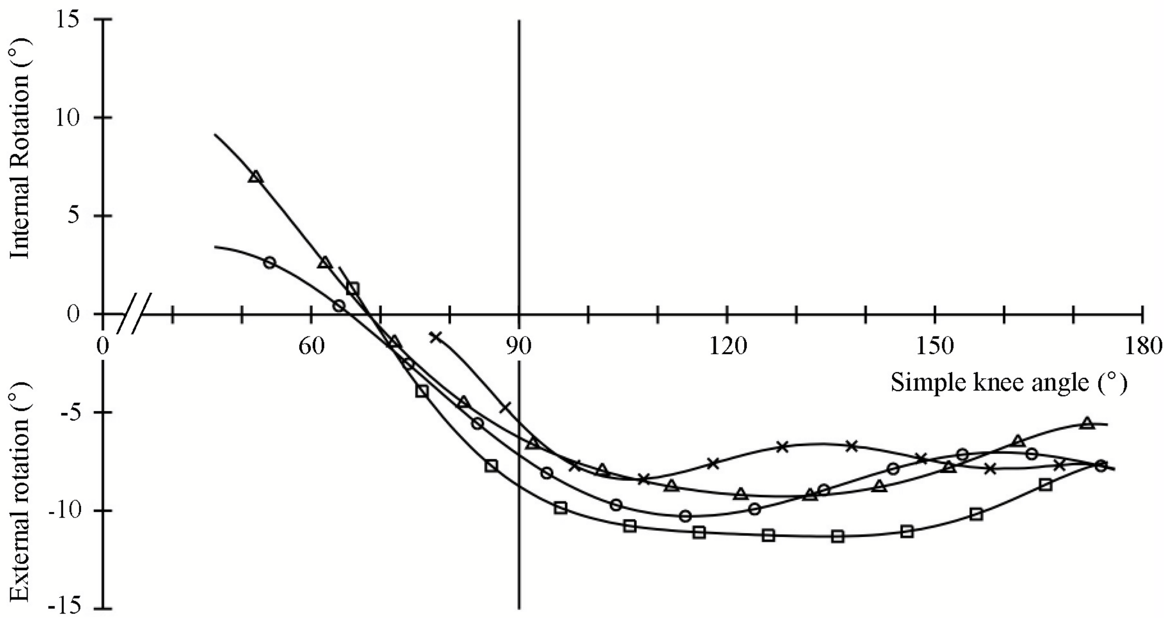

Figure 3. Polynomials of the γ versus κ relationship in the four unloaded trials of a typical subject.

The β values showed little variability (mean of the average standard deviations = 0.7 ± 0.5°; mean of the maximum standard deviations = 1.1 ± 0.7°). However, the γ values showed much larger variability (mean of the average standard deviations = 3.1 ± 2.3°; mean of the maximum standard deviations = 4.9 ± 2.8°). These results indicate that it is possible to predict reliably the unloaded value of β from the κ angles of any loaded activity, but not the unloaded value of γ. Because of this, no attempt was made to use unloaded γ angles from the value of κ to predict internal/external rotation in the loaded trials.

Figure 4 shows the β values of a typical cutting trial, the predicted unloaded β values for each instant of the loaded trial based on the κ angles of the loaded trial, and the adjusted β angles that resulted from subtracting the latter from the former. The β angles of the unloaded trials has some tendency to follow the β angles of the loaded trials. Therefore, subtracting the β angles of

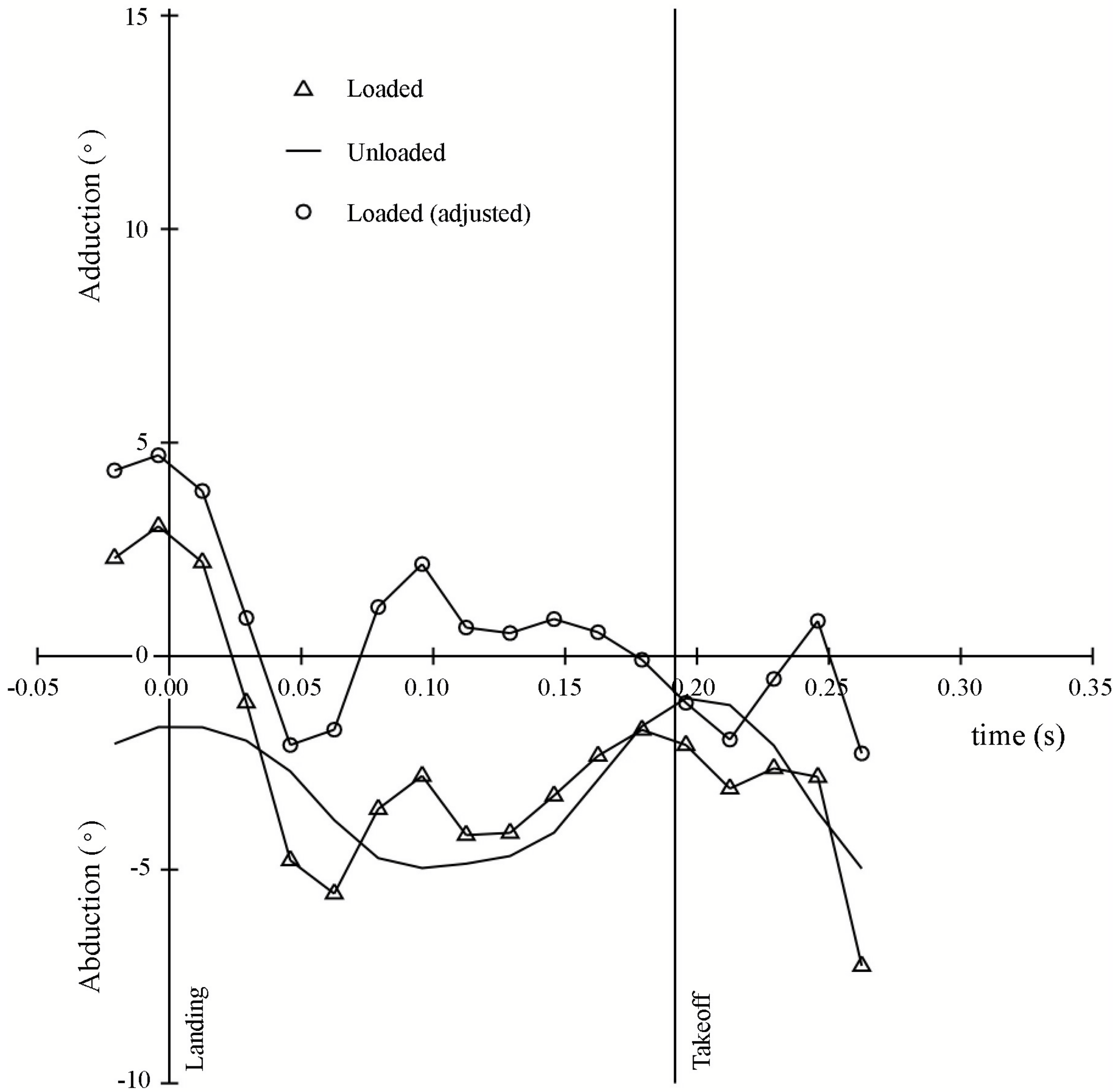

Figure 4. β values of a typical straight trial, predicted unloaded β values for each instant of the loaded trial based on the κ angles of the loaded trial, and adjusted β angles that resulted from subtracting the latter from the former

the unloaded trials from the β angles of the loaded trials tended to produce adjusted loaded β angles that had smaller magnitudes than the unadjusted loaded angles. This implied that the unadjusted β angles tended to overestimate the magnitude of the ab/adduction deformation of the knee in the loaded trials. The average of the absolute values of the loaded β angles was 4.1 ± 2.4°; the average of the absolute values of the unloaded β angles was 3.3 ± 1.9°. The result of subtracting the predicted unloaded values from the loaded values was a reduction in the magnitude of the adjusted β angles to an average absolute value of 2.3 ± 1.4°. A paired t-test indicated that this change was statistically significant ($P < 0.003$). Figure 4 shows some of the benefits of using the β value adjustment. According to the unadjusted β values, the knee of the subject shown in Figure 4 would be in adduction, whereas after making the adjustment it is clear that the knee deformation was generally in the direction of abduction.

Figure 5 shows the β values of a typical straight run trial, the predicted unloaded β values for each instant of the loaded trial based on the κ angles of the loaded trial, and the adjusted β angles that resulted from subtracting the latter from the former. As in the cutting trials, in the running trials the β angles of the unloaded trials had some tendency to follow the β angles of the loaded trials. Therefore, subtracting the β angles of the unloaded trials from the β angles of the loaded trials tended again to produce adjusted loaded β angles that had smaller magnitudes than the unadjusted loaded angles. This implied that the unadjusted β angles tended to overestimate the magnitude of the ab/adduction deformation of the knee in the loaded trials. The average of the absolute values of the loaded β angles for the straight runs was 3.9 ± 1.4°; the average of the absolute values of the unloaded β angles was 3.3 ± 1.8°. The result of subtracting the predicted unloaded values from the loaded values was a reduction in the magnitude of the adjusted β angles to an average absolute value of 2.2 ± 0.7°. A paired t-test indicated that this change was statistically significant ($P < 0.013$). Similar to what was shown for the cutting trials in Figure 4,

Figure 5 shows some of the benefits of using the β value adjustment for the straight runs. According to the unadjusted β values, the knee of the subject shown in Figure 5 would be in

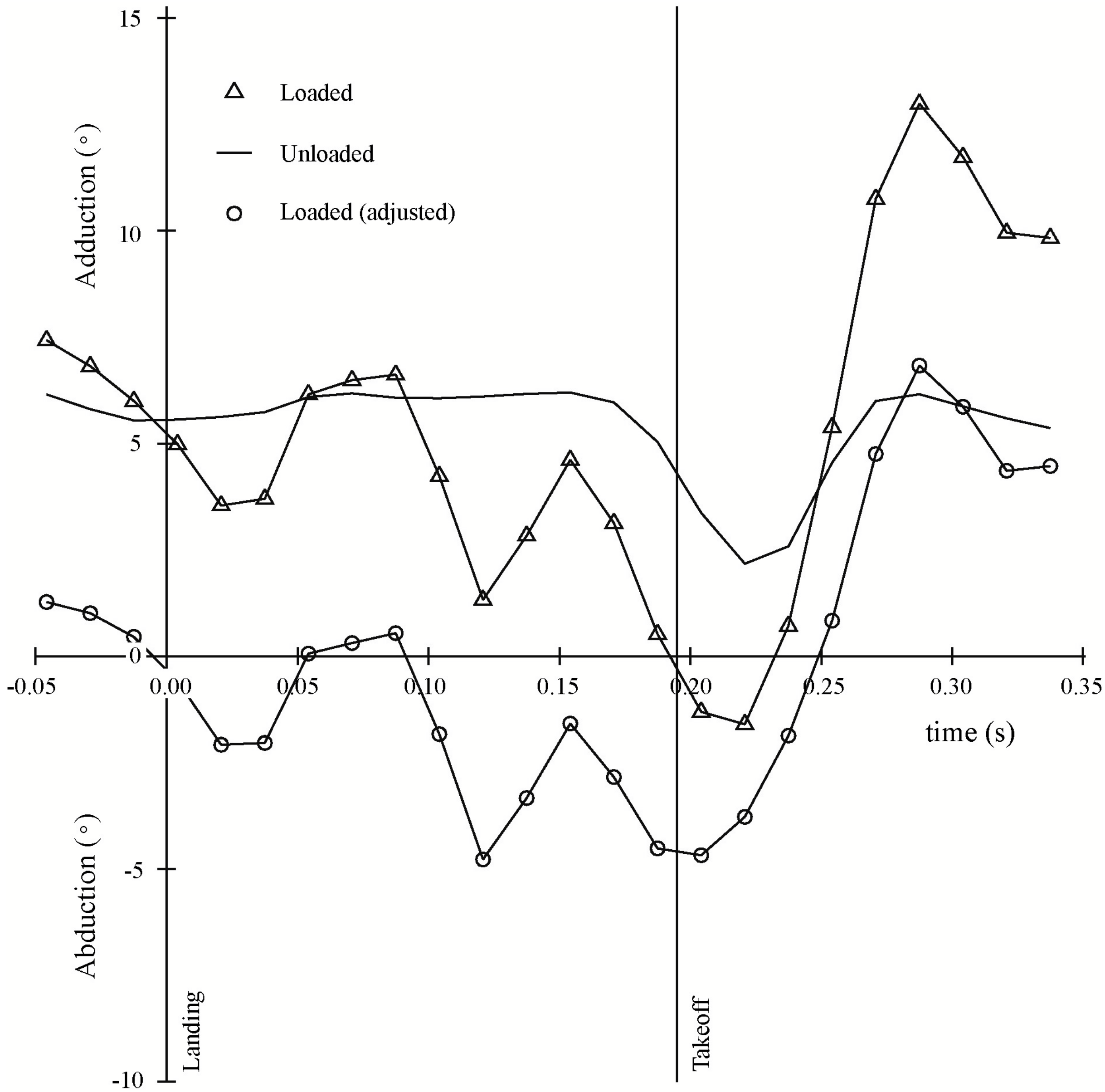

Figure 5. β values of a typical cutting trial, predicted unloaded β values for each instant of the loaded trial based on the κ angles of the loaded trial, and adjusted β angles that resulted from subtracting the latter from the former

abduction, whereas after making the adjustment it is clear that the knee deformation was close to neutral, with an average value that was slightly in the direction of adduction.

A comparison of Figures 4 and 5 shows that the unadjusted loaded β angles and the unloaded β angles had positive (i.e. adduction) values in the subject shown in Figure 4, and negative (i.e. abduction) values in the subject shown in Figure 5. This could reflect structural differences between the knees of those two subjects. However it could also reflect accidental differences in the placement of the reflective markers, as will be shown by the following simple example. Let us consider a hypothetical shank orientation that consists of an α rotation of -90° and a β rotation of 0° when the knee markers are placed in the correct positions. If the knee markers were placed accidentally at a 1° externally rotated orientation relative to where they should have been, the α rotation would be about this externally rotated axis. After -90° of α rotation, a β rotation of -1° would be needed to bring the longitudinal axis of the shank to its correct position. Conversely, accidental placement of the knee markers at a 1° internally rotated orientation relative to where they should have been, would require a β rotation of 1°to bring the longitudinal axis of the shank to its correct position.

CONCLUSIONS

The β Cardan angles of the unloaded knee have quite consistent values within each individual subject, but the γ Cardan angles do not. Because of this, the deformation of the knee joint in the direction of ab/adduction in the course of a loaded activity can be calculated quite effectively by subtracting the β angles of an unloaded motion from the β angles of the loaded activity. However, this approach is not feasible for the γ angles (internal/external rotation) due to the latter's much greater variability within subjects in the unloaded trials.

Our subjects were aware beforehand of which trials would be straight runs and which ones would require a cutting maneuver. As a result, they anticipated the cutting maneuvers, and began moving to the left before the designated support period. Their final direction of motion was also less leftward than planned. Therefore, the change in direction of motion during the analyzed support in the cutting trial was much smaller than planned. This may have been the reason why the average β values of the cutting trials were not very different from those of the straight runs.

The β angles of the unloaded motions had some tendency to follow patterns similar to those of the loaded activities. Therefore, the adjusted β angles generally reached less extreme values than the unadjusted ones. The standard Kadaba/Davis methods do not include such adjustments. Therefore, the ab/adduction angles calculated with these methods are likely to be inflated.

The results of the present project support the measurement of angles in unloaded trials to provide adjustments for the raw β angles of loaded activities. On the other hand, such an approach is not currently possible for the γ angles due to the large intra-subject variability of the γ angles in the unloaded trials.

Future work in this area should concentrate on the measurement of more accurate angle values through the use of more modern equipment that will produce more accurate coordinate values as well as a larger frequency in the data collection technology.

In summary, the present project demonstrated a new method for improving the calculation of knee configuration angles. These results can be applied in both clinical and sport biomechanics in ways that will ultimately be of benefit in the future study and treatment of knee injuries.

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Thesis: “A method to improve the calculation of knee configuration angles in clinical and sport biomechanics”

Area of Interest: Computer Science

Bachelor of Science in Education: Department of Physical Education and Sport Science, Zinman College, Israel. October 2003 to July 2007.

Major: Posture Cultivation (Advisor: Dr. Vardita Gur)

Support Area: Biomechanics

Area of Interest: Sport injuries

Professional Experience

Associate Instructor, 2009-2010

Biomechanics Laboratory, Indiana University

Military Service

Engineering Division, Israel Defense Forces, November 2000 to November 2002.

Fitness Trainer, Israel Defense Forces, November 2002 to November 2003.

Languages

Hebrew – Native Language

English – Fluent

French – Basic

Arabic - Basic